Fundamentals of Physiologic Monitoring

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For centuries, medical practitioners had no electronic medical instruments and had to rely on their senses of sight, hearing, smell, taste, and touch to obtain physiologic measurements. Although it is possible to estimate blood pressure by palpating the pulse at the radial or brachial artery, such estimates are not accurate. Determining arterial oxygen saturation of hemoglobin is more complex: how "blue" a patient appears depends on skin coloration, lighting, and the examiner's sense of color. Finally, using radiographic images to validate pulmonary edema when clinicians suspect that there is an elevated left atrial or pulmonary artery wedge pressure also challenges human senses. However, today's medical instruments use transducers and signal processors to convert patient information into a form that clinicians can easily perceive and understand. This article defines terms used with biomedical instrumentation and discusses the components of ideal physiologic patient monitoring systems (KEYWORDS: physiologic monitoring, transducer, instrumentation).

MONITORING SYSTEMS

Figure 1 shows a block diagram of a generalized physiologic instrumentation system. It is composed of multiple components. Primary information flow is from left to right. The physical quantity measured, such as an electrocardiogram (ECG) or a pressure waveform, is termed the "signal."

Transducers

Transducers convert signals from one energy form to another. The final signal usually is electrical because the technology for amplifying and displaying electrical signals is so well developed. The ideal transducer should:

respond only to the signal it was designed to detect (such as pressure) and ignore all other signals (such as temperature or light), and

connect to patients in a way that is minimally invasive and by which the energy extracted is minimized (i.e., if a patient's blood pressure dropped 25 mmHg when an arterial catheter was inserted, the devices would not be used).

Transducers can be as simple as ECG electrodes. However, other transducers may require an energy source, such as the DC voltage used to power a pressure transducer or pulses of electrical energy that activate the light sources in a pulse oximeter.

Amplifier and Signal Conditioning

Seldom can transducer output signals be directly attached to a display device. Usually the signal amplitude from transducers is small (typically in microvolts) and must be amplified and filtered. Adjusting the amplification is much like adjusting the volume on a

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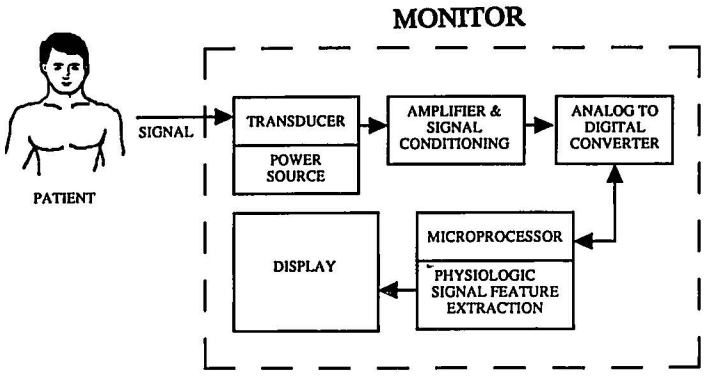

Figure 1. Diagram of a generalized instrumentation system. The transducer converts energy or information from the signal to another form of energy, usually electrical. The signal is then processed, features extracted such as systolic and diastolic pressure, and displayed by the monitor so humans can view and act on the information.

radio or stereo. Signal conditioning includes filtering the signal to rid it of high or low frequency noise. For example, with ECG signals, a high-pass filter is used to eliminate low-frequency signals, thereby minimizing artifact caused by patient movement. Direct pressure measurements require keeping the low-frequency signal, which is the mean pressure. Because each pressure transducer may have different voltage outputs with no pressure applied, the monitoring system must be capable of compensating for this offset voltage.

Analog-to-Digital Conversion

Today most medical devices use only simple amplifiers and filters and quickly convert the derived electrical signal into a digital form using an analog-to-digital converter (ADC). These convert "analog" signals, such as an ECG, to digital form, usually with 10-bit resolution (2^{10} = 1,024 or a 1 part in 1,024 resolution) at rates of 100 to 300 times per second.

Microprocessor

Signal processing and display usually are accomplished with inexpensive and reliable microprocessors. These take signals, such as an arterial pressure waveform, and derive and display heart rate, systolic, diastolic, and mean blood pressure in real time. Reliable and representative physiologic signal feature extraction is one of the remarkable enhancements made available by monitoring systems. These computers also are used to compensate for undesirable transducer characteristics such as nonlinearities and to average repetitive signals to reduce unwanted noise.

Display

Results of signal acquisition from transducers, their amplification, processing, and feature extractions are displayed in an easy-to-understand format. The display device usually is a video screen, but in recent years liquid crystal displays (gray), light-emitting diodes (red), and gas-plasma displays (orange) have come into common use.

TRANSDUCERS

It seems that each year new methods and technologies are developed for measuring parameters that previously had been thought impossible or improbable to measure. The number and type of transducers used in medicine, especially for physiologic monitoring, are large and diversified. Bioinstrumentation texts[1,2] and encyclopedias[3] provide detailed information about transducers.

Pressure Transducers

Modern blood pressure transducers have resulted from a remarkable set of advances in technology.[4-6] Ten years ago, blood pressure transducers were "reusable" devices that cost about $500 each. These devices required handling with great care, they had substantial

zero drift and calibration errors, and they often failed. In addition, their high cost mandated they be reused, which placed the patient at risk for infection. This threat led to the use of "disposable domes" for coupling the fluid-filled catheter/tubing system to the transducer. These domes caused additional degradation in accuracy and dynamic fidelity of the pressure monitoring system. Today semiconductor technology has produced the transistor, processor chips for personal computers, and digital watches. The same chip technology now is used to make blood pressure transducers that can be purchased for less than $20 and thus are disposable. Disposable transducers are much more accurate, rugged, stable, and inexpensive, when compared with reusable transducers. Because it is impossible to thoroughly clean and sterilize disposable transducers, they should *not* be reused.

Figure 2 shows a 10-cm diameter silicone semiconductor wafer. This wafer is only 0.40 mm thick and contains about 600 operational blood pressure transducers. Each of these 2.5-mm by 2.5-mm transducer elements is a chip. Figure 3 shows a cross-sectional and top view of a transducer chip. Figure 4 is a close-up of a transducer chip mounted on its plastic holder. The fine, gold wires that attach the transducer leads to the chip can be seen.

The semiconductor chip performs like most other pressure transducers. It contains a diaphragm that deflects or bends (Fig. 5) when pressure is applied. Figure 5 illustrates (in exaggerated dimensions), how the diaphragm deflects when pressure is applied. Volume displacement of blood pressure transducers is about 0.1 mm^3 per 100 mmHg pressure applied. Thus, the approximate displacement for a 2.5-mm by 2.5-mm transducer is 0.16 μm for each mmHg applied and 16 μm for each 100 mmHg applied. This small displacement is about one third the diameter of a human hair (50 μm). Pressure transducers for measuring blood pressure have been standardized by the American National Standards Institute (ANSI)[4] and have sensitivities of 5 microvolts per volt of excitation per mmHg pressure applied. Thus, if 5 volts DC is applied, an output of 25 microvolts per mmHg pressure applied is generated; this is a small signal. The new state-of-the-art transducers are rugged, have low temperature and time drift, and are accurately calibrated to within (±2% or ±1 mmHg, whichever is larger).

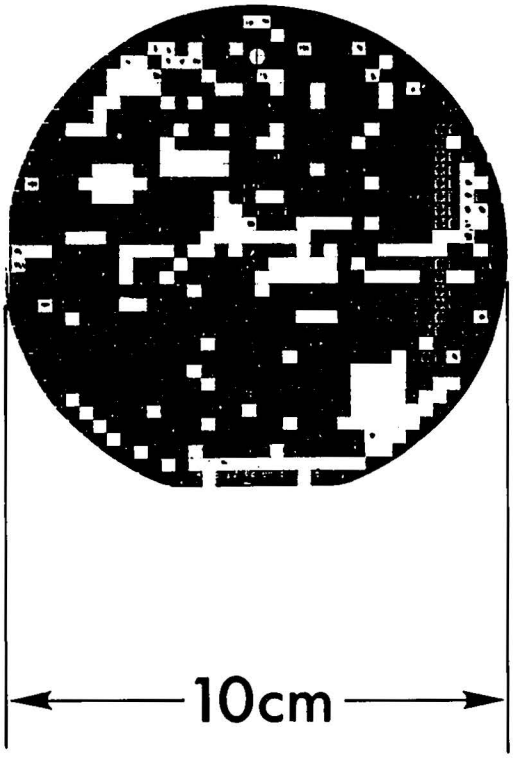

Figure 2. **Top view of a 10-cm diameter silicone semiconductor wafer (0.40 mm thick) with 557 2.5 × 2.5-mm medical pressure transducer chips. Where there are open segments, the transducers in this simple chip form have been tested, found to be defective, and discarded.**

Temperature Transducers

Measurement of body temperature has been an important part of health care for centuries. Today it remains a reliable parameter when measured in combination with other indicators to assess the state of health.[7] Each special application for monitoring temperature may have a unique need. For example, the classic mercury-in-glass thermometer may perform adequately for an occasional measurement of body temperature but is inappropriate if instrument size or speed of response is important or breakage is a problem. Thermistor sensors are convenient electronic temperature transducers that are small and provide a fast response at relatively low cost.

A thermistor provides small size, high reso-

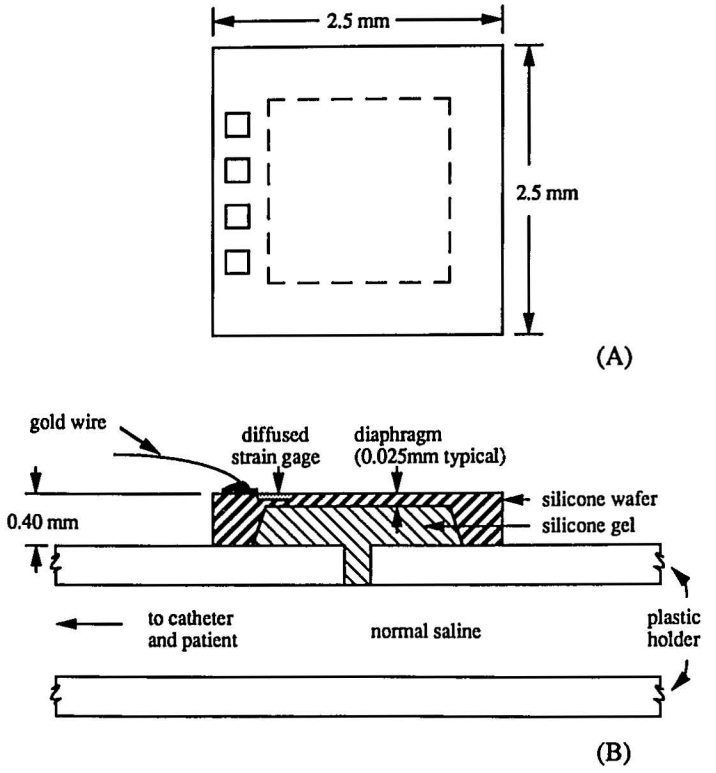

Figure 3. Top and side views of a single silicone semiconductor pressure transducer. The top view (A) shows the four wiring pads used to connect the diffused strain gage of the chip to the cabling that goes to the bedside monitor. The dashed line indicates the diaphragm area of the transducer that had been etched out of the back side of the chip. The side view (B) shows the thin pressure sensing diaphragm (0.025 mm) and how the chip is attached to the plastic transducer holder. The patient's pressure is coupled to the diaphragm via the saline column and the silicone gel as indicated. (Dimensions are not drawn to scale.)

lution, and rapid response for blood temperature measurement at a point along an indwelling catheter. As an example, a thermistor sensor for thermodilution cardiac output monitoring with pulmonary artery catheters is fabricated by forming a powdered semiconductor material (usually a metal oxide) into a small bead around two lead wires or by melting the semiconductor into a pellet shape and coating its opposite faces with conducting electrodes.[7] The resistance of the resulting thermistor has large resistance variation temperature. Unfortunately, the resistance-temperature curve is nonlinear. However, unique and stable characteristics of thermistor transducers are easily determined. Thermistors can be small (0.2 mm diameter); they have response times of less than 0.1 second and temperature ranges of 100°C; and they can be trimmed to interchange within ±0.1°C of one another.

Oximetry

Oxygen saturation measurement relies on the difference in optical absorption spectra of hemoglobin and oxyhemoglobin. As a result, various optical methods for measuring oxygen saturation have been developed by using two wavelengths in the red/infrared region. The red wavelength (about 660 nanometers) has the largest difference in light absorption between hemoglobin and oxyhemoglobin; at the infrared wavelength of 805 nanometers the two hemoglobins have the same absorbance. The 805-nanometer wavelength is called the "isosbestic" wavelength.

Pulse Oximeters. A recent and major advancement in oximetry has been made with the pulse oximeter.[8-10] Contemporary pulse oximeters use two light-emitting diodes to shine light through a variety of body locations (fingers, ears, or nose), where a small photodiode detects the transmitted signal. These inexpensive disposable probes and their associated processing and display capability have revolutionized oxygen saturation monitoring.

Light-emitting diodes are ubiquitous. Originally used in digital watches, they now frequently appear as displays on intravenous

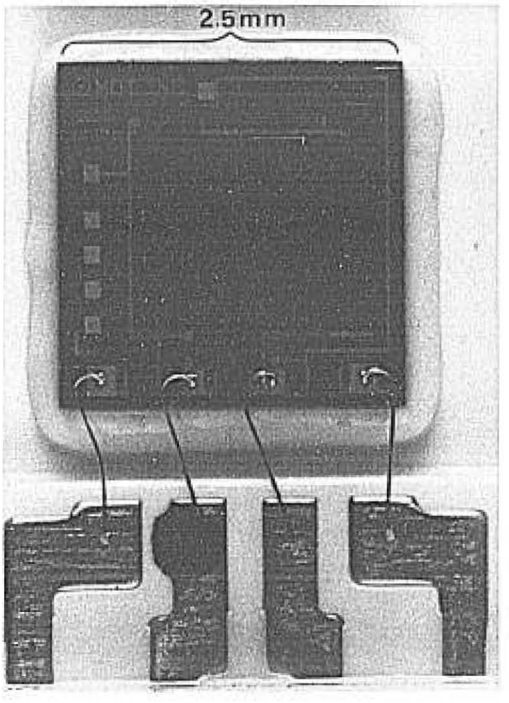

Figure 4. Closeup of the top side of a partially assembled transducer chip mounted on its plastic holder. Note the small gold wires attached to the pads on the chip and bringing the leads to the outside, where they will connect by a cable to the patient monitor.

fluid controllers and pumps. The receiving photodiode detects pulsating light from the red and infrared light sources after they have been transmitted and absorbed by the tissue. When the pulsatile components of the light are less than 0.5% of the steady (DC) component, the system's accuracy falls. It is possible to measure oxygen saturation noninvasively between 50% to 100% with an acceptable clinical accuracy of about 2.5%. Pulse oximeters do not measure other types of hemoglobin, such as carboxyhemoglobin or methemoglobin. Pulse oximetry has revitalized oximetry, which reverses a trend that began when the oxygen electrode became available and partial pressure of oxygen measurements became feasible. Severinghaus and Honda said, "Pulse oximetry is arguably the most significant technologic advance ever made in monitoring the well-being and safety of patients during anesthesia, recovery, and critical care."[10]

Fiberoptic Reflection Oximeters. The first report of oxygen saturation measurement by reflected light from blood was described in 1949.[8,11,12] It was not until fiberoptic catheters and inexpensive light sources and detectors became available that this mode of measuring oxygen saturation became widely used. Mixed venous oxygen saturation measurements are routinely taken for selected patients with disposable fiberoptic pulmonary artery catheters that use this technology.

CHARACTERIZATION OF INSTRUMENTATION SYSTEMS

Just as critical care has its own vocabulary, so does measurement and instrumentation technology. To assist the reader, several important measurement terms are discussed and defined. The Appendix provides a glossary of terms. One of the primary concerns when making a measurement is to know its accuracy.

Accuracy of a measured quantity is the true value minus the measured value divided by the true value (usually expressed as a percentage). True values are seldom known exactly, but a reference value that is traceable to the National Institute of Standards and Technology (NIST) usually is used.

Precision of a measurement expresses the number of distinguishable alternatives from which a given result is selected. A weight of 75.2 kg is more precise than a weight of 75 kg. High precision does not imply high accuracy.

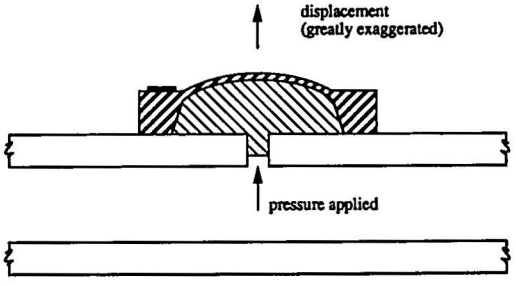

Figure 5. Diagram of the side of a transducer chip with positive pressure applied showing its diaphragm displacement (exaggerated). The diaphragm of the silicone chip transducer is much like a large glass window with the wind blowing against it. It moves, but slightly.

Resolution is the smallest incremental quantity that can be measured with certainty. For example, for measuring height the smallest scale may be 1 cm.

Reproducibility is the ability of an instrument to provide the same output for an equal input. Slightly different results may be obtained when the same weight is placed on a scale several times. The reproducibility of transducers and measuring systems typically is much better than their accuracy. For example, a pressure transducer may have 2% accuracy but may make repeated measurements of the same pressure to within 0.5%.

Linearity for transducers is the extent to which the output signal is proportional to the applied input signal. For an amplifier, linearity is the extent to which amplification is accomplished without amplitude distortion. Figure 6 shows the input/output characteristics of a "linear" and "nonlinear" system.

Sensitivity, gain, or *amplification.* Sensitivity usually defines the relationship between transducer input and output. A highly sensitive transducer is one that produces large output from a small input signal. Occasionally, the amplification or gain of an amplifier also is referred to as sensitivity. Amplifiers have high sensitivity if they have high gain or amplification. In Figure 6, the sensitivity is shown graphically as the slope of the input-output characteristic curve.

Calibration refers to the process of adjusting the sensitivity of a transducer or adjusting the gain of the amplifier. Figure 7 shows the effect of a 15% sensitivity increase that may have resulted from an increase in transducer sensitivity or an increased amplifier gain. For physiologic signals such as pressure waveforms it is desirable to have fixed and stable *transducer sensitivities* and *amplifier gains.* Modern disposable pressure transducers and monitors have sensitivities and gains that are so stable that it takes testing equipment and methods of standards laboratory grade to validate the sensitivities and gains.

Zero/offset refers to a transducer or amplifier condition for which the "0" input signal is an output signal called the "zero offset" (Fig. 8 shows a pressure signal with a −11 mmHg pressure offset). Offset can be positive or negative, and for pressure monitoring systems often is caused by the position of the "zeroing" stopcock in relation to the transducer.

Hydrostatic effect is the pressure applied on the transducer as a result of fluid column height above or below the transducer. Hydrostatic pressure can be positive or negative, depending on where the transducer is in rela-

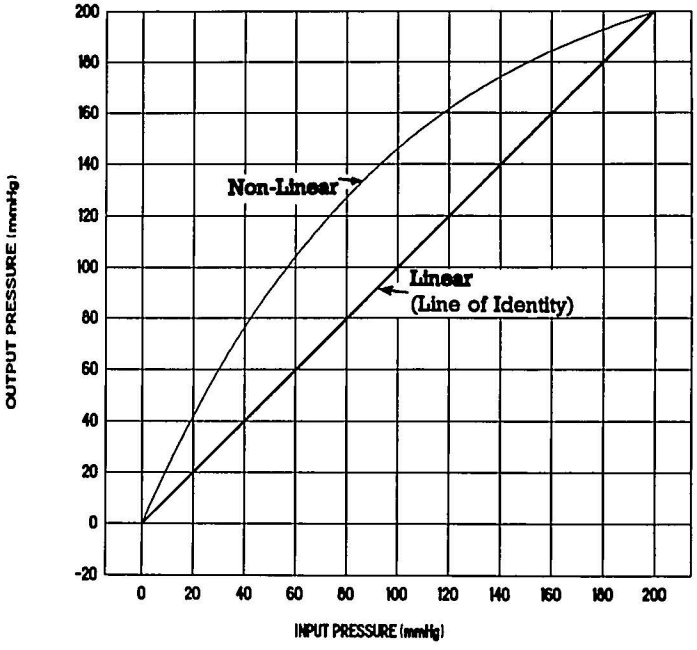

Figure 6. The X axis is input pressure and the Y axis is output pressure. Note, for the Linear system, the output pressure corresponds exactly to the input pressure and forms the line of identity. On the other hand, the output from the non-linear system agrees with the input pressure only at 0 and 200 mmHg. At 30 mmHg input pressure, the output pressure is 60 mmHg.

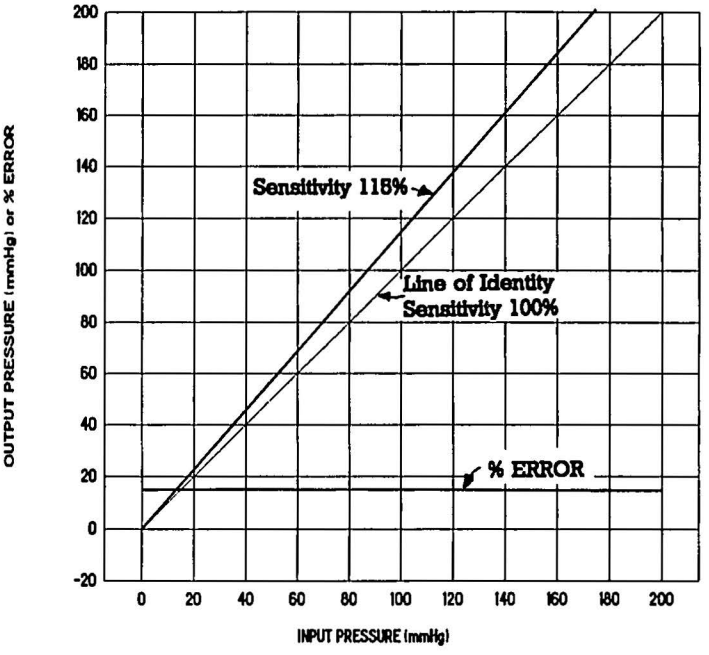

Figure 7. The characteristics of a transducer that has a 15% increased sensitivity or an amplifier system with a 15% increase in gain. In either case, the monitoring system has a 15% sensitivity increase. Note that there is a 15% output pressure error over the entire input pressure range.

tionship to the midaxillary line. Mercury is 13.6 times more dense than water (normal saline). Thus, as Figure 9 shows, for each 13.6 cm of fluid height, there is a 10 mmHg pressure change.

Static response. Medical instrumentation has static (forces at rest or in equilibrium) and dynamic requirements. Static requirements usually refer to the performance of an instrument to low-frequency events, such as

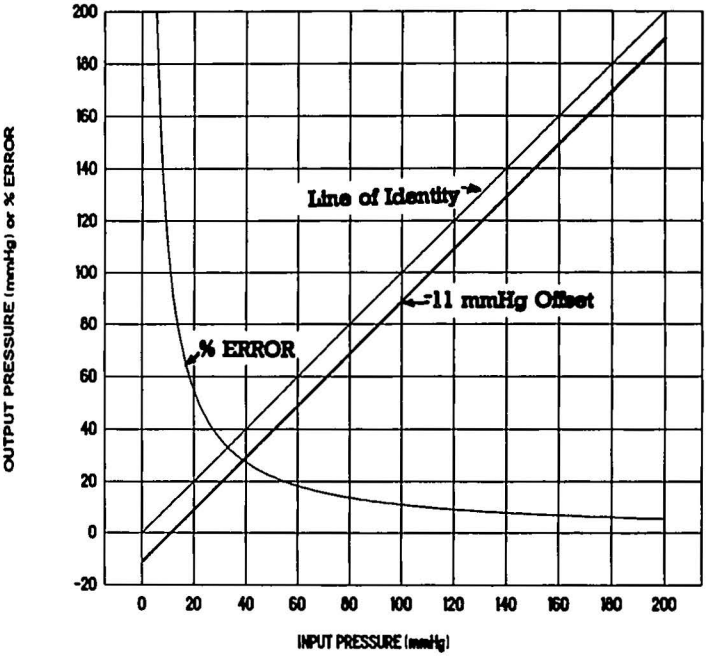

Figure 8. Shows the effect of an 11-mmHg "zeroing" error. (For a clinical example, see Fig. 12.) Over the entire input pressure range, there is an 11-mmHg under-reading or negative offset. Also shown is a percentage error curve. In the low pressure range (typical of venous and pulmonary artery pressures) the errors are enormous. For example, at an input pressure of 11 mmHg, there is a 100% error.

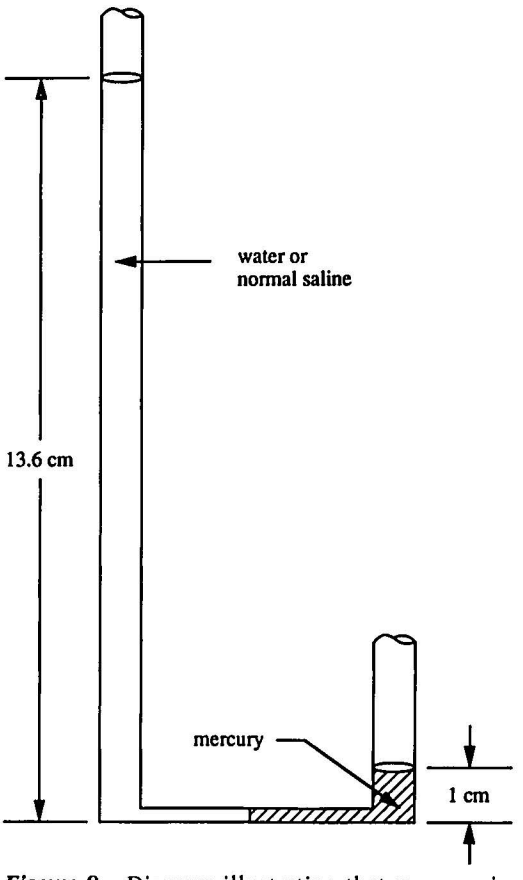

Figure 9. Diagram illustrating that mercury is 13.6 times more dense than water. A 13.6-cm column of water causes the same pressure as a 1-cm column of mercury. Stated another way, each vertical 1.36 cm of water is equivalent to 1 mmHg pressure.

measurement of human weight, which normally changes slowly. Dynamic requirements refer to rapidly changing measures that vary at high frequency, such as the arterial blood pressure waveform.

Dynamic response refers to how well the instrumentation system can reproduce part of the signal that changes with time. For example, with the ECG and blood pressure signals, the dynamic response is the "pulsatile" part of the waveform. It is important for the instrumentation system to provide high quality reproduction of the pulsatile and dynamic features of ECG signals and blood pressure waveforms.

PRESSURE MONITORING SYSTEMS

Figure 10 is a schematic diagram of a measurement system used for determining arterial blood pressure. An arterial catheter is placed in the radial artery. The catheter is connected through a piece of plastic tubing to stopcock 1, which is connected to stopcock 2 through a segment of pressure tubing, where it attaches to a pressure transducer and to a continuous flush device. The transducer is connected to a monitor (Fig. 1) that displays a pressure waveform and derived parameters.

Setting the Zero Point for the Monitoring System

Figure 11 shows the proper technique for setting the zero point for a pressure monitoring system with the stopcock opened to air at the midheart level (midaxillary line). Care should be taken to determine the midaxillary or phlebostatic reference point accurately.[13,14] Figure 12 shows an improper setting of the zero point of a pressure monitoring system by opening the stopcock when it is 15 cm above the midaxillary line. In this case, the pressures measured from the pulmonary artery catheter would be reported as being 11 mmHg too low because of the *hydrostatic effect.* If the zeroing point had been 15 cm below the midaxillary line, the pulmonary artery pressure would have been overestimated by 11 mmHg. Setting the zero point, or "zeroing," the pressure monitoring system is the *single* most important step in setting up a pressure measurement system. Zeroing causes the largest pressure measurement errors in the clinical situation. Transducer and amplifier sensitivities usually are accurate to within 2%. However, large percentage pressure errors can occur because of the improper zeroing of pressure monitoring systems, especially when measuring venous and pulmonary artery pressures because these pressures are small. For example a zeroing misalignment such as that shown in Figure 12 results in a 11 mmHg error. If the patient had a pulmonary artery wedge pressure of 35 mmHg, only 24 mmHg would have been measured, representing a 31% error. The

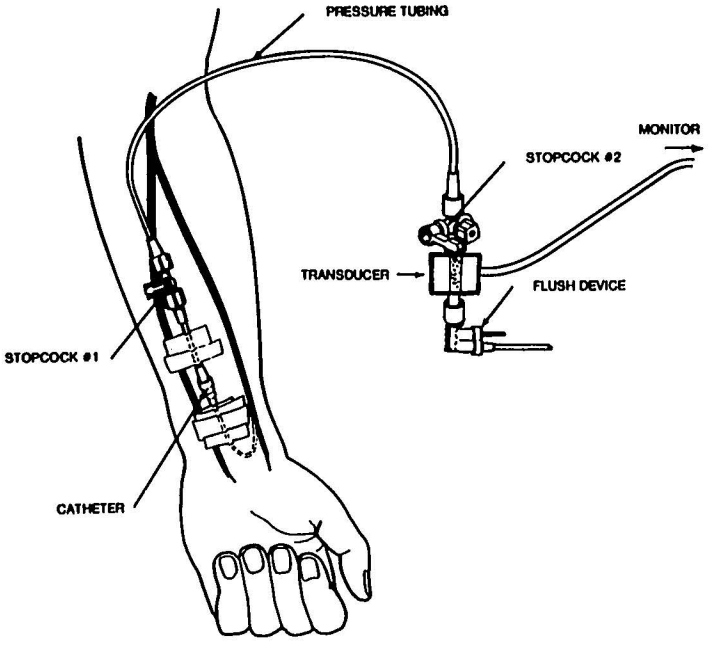

Figure 10. Catheter-tubing-transducer-flush device pressure monitoring "plumbing" system schematic. The catheter in this case is in the radial artery. Stopcock #1 is used as a site for withdrawal of blood samples. When fluid filling the pressure monitoring system, care should be taken to fill all the central cavities of the stopcock to eliminate trapped air bubbles. Stopcocks are a hazard for patient contamination and infection, so that ports should be covered with sterile caps. The pressure tubing is used to position the transducer in a convenient location. The tubing used should be noncompliant to assure high-fidelity transmission of the patient's pressure waveform to the transducer. The transducer converts the pressure variations in the plumbing system to an electrical signal that can be analyzed and displayed by the bedside monitor. The continuous-flush device helps prevent blood from accumulating and clotting in the catheter tip. The fast-flush valve is used to test the system's dynamic response characteristics.

magnitude of an 11 mmHg zeroing error is shown graphically in Figure 6.

Many pressure monitoring users erroneously assume that the location of the pressure transducer is the zero reference point. The only time such an assumption is correct occurs when the stopcock is attached to the transducer and is at the same elevation as the

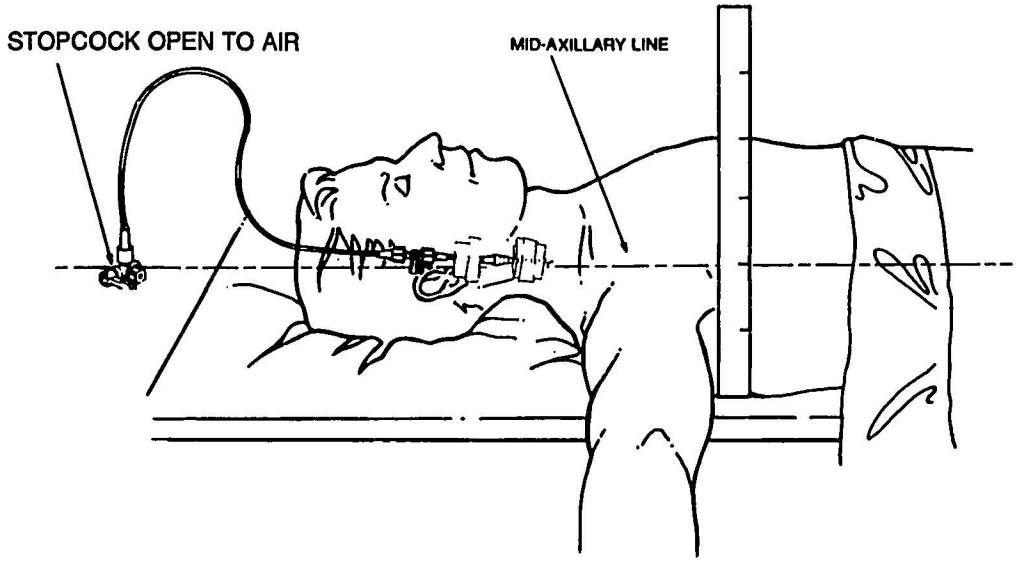

Figure 11. Proper zeroing of a pressure monitoring system. The air-water interface of the zeroing stopcock is placed at the mid-heart or mid-axillary line. (Adapted with permission.[15])

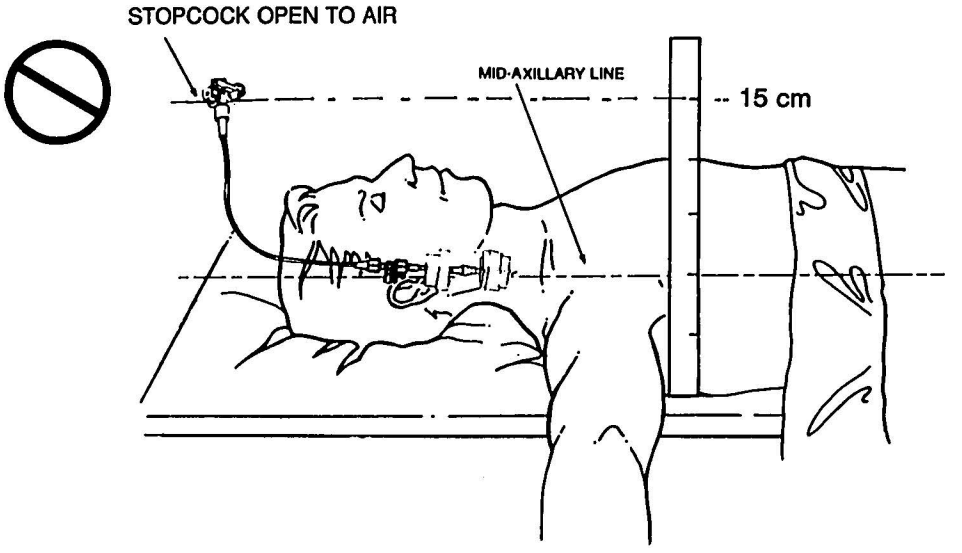

Figure 12. **Improper zeroing of a pressure monitoring system. The air-water interface of the zeroing stopcock is placed 15 cm above the mid-heart or mid-axillary line. If the pressure monitoring system is zeroed with this approach, all pressures measured will be under-stated by about 11 mmHg (15/1.36). (Adapted with permission.[15])**

transducer. The proper zeroing reference set-up is made by aligning the open stopcock port to air at the midaxillary line. Pressure monitoring systems should be "zeroed" frequently and always before treatment changes are initiated based on pressure data. Zero changes can be caused by many factors: 1) patient's vertical position changes in relationship to the pressure transducer (hydrostatic effect), 2) transducer zero drift caused by changes that occur with temperature or time (Fig. 8), and 3) pressure amplifier drift (Fig. 8).

Calibration of System Sensitivity

The sensitivity accuracy of a standardized pressure transducer is fixed by the manufacturer to be 5.0 microvolts per volt of excitation for each mmHg of pressure applied (5.0 μV/V/mmHg) to within ±1%. This degree of accuracy is adequate for most clinical monitoring purposes. Thus, using the new standardized disposable transducers, only zeroing is required. If there is concern that a transducer or monitor may have an inappropriate calibration, connecting a 30-cm vertical sterile fluid column should yield a pressure of 22 mmHg. If a mercury column is used to perform the check of calibration, extreme care should be used to assure that the fluid-air interface does not cause a break in sterile technique and that the person performing the procedure does not turn the stopcocks inappropriately because such an error could introduce an air embolism into the patient.

Dynamic Response Testing

In the intensive care setting, the catheter-tubing-transducer system usually can be characterized as an underdamped second-order system analogous to a bouncing basketball or a ringing bell. Once a basketball is dropped, it bounces several times and comes to rest.[15-18] If the ball is not properly inflated, it will come to rest with fewer and longer intervals between bounces. The same analogy applies to a blood pressure monitor system. If the system has low volume displacement, it will oscillate rapidly and come to rest when it is perturbed by activating pressure from the fast flush valve. Such a system will have a natural frequency and a damping coefficient. If the system has air bubbles or the tubing is long and compliant (for example, if venous

tubing is used instead of pressure tubing), the system will be underdamped, and the systolic pressure will be underestimated and diastolic pressure overestimated. Only the mean pressure will be accurate.

Releasing the fast flush valve and allowing the valve to close generates a step change in pressure, much like dropping a basketball. Once the fast-flush test has been executed, it is possible to quickly and easily determine the dynamic characteristics of the system. (Waveform details and testing methods may be found elsewhere in this symposium.)

Optimizing Dynamic Response.

To optimize dynamic response, one must follow the procedures described here.

Eliminate all air bubbles in the system. This is the single most important step used to optimize dynamic response. Common locations of air bubbles are in stopcocks, where all "ports" have not been filled with fluid, and at luer-lock interconnect points. Light tapping while fluid is filling the system is an effective method for removing air.

Removing air from above the flush solution bag prevents air from "going into solution" at high pressure, then at a lower pressure (near the catheter or low flow locations) coming out of solution and forming an air bubble.

Use short lengths of pressure tubing.

Mount transducer on patient if possible.

Use low compliance (hard) pressure tubing (not distensible venotubing).

Minimize the number of connections.

Use only low compliance pressure transducers.

Heparin Use and Intermittent Versus Continuous Flushing

Heparinized saline has long been used to maintain the patency of indwelling vascular catheters.[19] Use of heparin in a continuous flush solution raises three questions:

1. Is a heparinized solution effective in maintaining catheter patency?
2. If it is effective, is the heparin adsorbed to the plastic infusion bag and tubing so that the concentration becomes diminished and ineffective over time?
3. How can heparin in the catheter be prevented from disturbing coagulation studies when samples are withdrawn from the catheter?

In recent years each of these issues has been addressed. Heparin flushing effectiveness was studied by Hook et al.[20] in 1987 and Clifton et al. in 1991.[21] These studies suggest that catheters can be maintained with intermittent flushing but that continuously flushing the catheter with a heparinized solution is associated with a decreased frequency of catheter occlusion and malfunction. However, in a recent study based on literature review, Peterson and Kirchoff[22] challenge the need to use heparin in the flush solution. In some unreported work done in 1980, we studied the concentration of heparin in an infusion bag. During a 24-hour period, we detected no decrease in heparin concentration. Thus, it appears that there is *not* a great amount of heparin absorbed into the walls of the plastic bags and tubing. Finally, Gregersen et al.[23] found that 5.1 mL of discard volume withdrawal was needed before the sample obtained reflected the patient's activated partial thromboplastin time. As a consequence, Gregersen et al.[23] recommended drawing activated partial thromboplastin time samples from a venous catheter.

CONCLUSIONS

Monitoring equipment that is properly understood, set up, and used provides nurses and other members of the patient care team with an ever-increasing amount of valuable data. Extreme care must be taken with all transducers and monitors to be certain that only accurate data are acquired. In most instances, care and attention to detail provide accurate acquisition of physiologic data. A troubleshooting guide for pressure monitoring systems is included (Table 1) to assist clinical staff in the prevention and correction of problems with their pressure monitoring systems.

TABLE 1. Troubleshooting Guide for Pressure Monitoring Catheter Systems

Problem	*Cause*	*Prevent or Fix*
Blood backup in	Loose connection or leak in flush bag pressure system	Return stopcock to closed position
catheter		Check luer-lock connections (may be loose)
transducer	Low flush bag pressure	Keep pressure bag inflated to 300 mmHg
flush device		
Air bubbles in	Inadequate flush system setup	Meticulous fluid setup (including bag) to avoid air bubbles
catheter	Improper stopcock position	Remove bubbles from all stopcock ports by gentle tapping and turning to all port positions
transducer	Leak or crack in catheter or flush device	Remove air bubbles from transducer
flush device		
Damped waveform with a poor fast flush (few oscillations)	Air bubbles	Remove air bubbles
	Leaks	Check connections for leaks
	Clotted catheter	Irrigate catheter with syringe
	No fluid in flush bag	Refill flush bag
	Low flush bag pressure	Repressurize flush bag
Display >200 mmHg when flush released	Broken flush device	Replace flush device
Pegging on bottom of display	Clot on catheter tip or catheter tip against vessel wall	Irrigate catheter with syringe
		Withdraw catheter 1 cm
Pressure <200 mmHg when flushing	Low flush bag pressure	Check flush bag pressure
Cannot aspirate blood	Clot on catheter tip or catheter tip against vessel wall	Aspirate clot if possible
		DO NOT force flush catheter, but if catheter irrigates easily, flush with syringe
		Withdraw catheter 1 cm
		Maintain heparin in flush solution if no clinical contraindication
Cannot irrigate catheter	Improper stopcock position	Check stopcock position
	Catheter clotted or kinked	Check catheter for kinks (see above for catheter irrigation)
Trace at top of display	Improper stopcock position	Check stopcock position (see above)
	Catheter clotted or kinked	
Cannot zero	Improper stopcock position	Check stopcock position
	Transducer bad	Replace transducer
	Bad monitor	Replace monitor
Kinked catheter	Acute catheter angulation	Check catheter and patient position to remove kinks
Broken connector	Misuse of catheter	Prevent blood loss
	Defective catheter	Prevent air embolism
	Stress on catheter hub	Replace catheter
No waveform on the display	Transducer disconnect	Connect transducer
	Monitor "off"/defective	Turn monitor on, replace if defective
	Bad "zero"	Rezero (see above)
	Catheter clotted or kinked	Replace transducer
	Defective transducer	

TABLE 1. Troubleshooting Guide for Pressure Monitoring Catheter Systems (*Continued*)

Problem	*Cause*	*Prevent or Fix*
Damp dressing around catheter site	Loose luer-lock connection	Check connections
	Cracked connection	Check catheter hub and catheter for cracks
Questionable pressure reading	Improper zero	Rezero and check transducer position
	Change in transducer position since zeroing	Check transducer calibration (see defective monitor and transducer procedures above)
	Transducer/monitor problems	

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APPENDIX

Glossary of Terminology

This glossary has been adapted from Webster.[3]

Accuracy. The accuracy of a single measured quantity is the true value minus the measured value, and this difference is divided by the true value; this usually is expressed as a percentage.

Example: Assume a perfectly set up arterial catheter monitoring system is in place and patient's measured systolic pressure is 120 mmHg. This measure becomes the *truth.* Then, the systolic pressure is measured with a cuff and the auscultatory technique and found to be 126 mmHg. The accuracy would be ([126 − 120] ÷120) = 5.0%. Properly established direct blood pressure measurement systems can have accuracies within ±3% of the measured value or ±1 mmHg, whichever is larger.

Precision. The precision of a measurement expresses the number of distinguishable alternatives from which a given result is selected.

Example: A digital scale that lists a weight as 56 kg has less precision than one that lists the weight as 56.51 kg.

Resolution. The smallest incremental quantity that can be measured with certainty.

Example: For most blood pressure monitoring systems, this is 1 mmHg. However, sometimes for venous water manometers, the resolution is 1 cm H_2O.

Reproducibility. The ability of an instrument to give the same output for equal inputs applied over a period of time. Also called "repeatability." Measurement systems usually have better reproducibility than accuracy.

Example: Assume the existence of a patient or patient pressure simulator that was exceedingly stable: the systolic pressure was always 120 mmHg. If multiple measurements of the pressure are taken with a catheter-manometer recording system, the values might be 120, 123, 119, 124, 120, 121, 122, 120, 118, and 120 mmHg. The mean systolic pressure is 120.70 mmHg; the standard deviation, a measure of reproducibility, is 1.83 mmHg; and the range of values from the "true" value is −2 to 4 mmHg.

Calibration. The adjustment of the designed operating characteristics; this usually is in comparison with a standard.

Example: To fix, check, or correct the graduations of a measuring instrument such as a thermometer.

Pressure. Defined as the force per unit area: Pressure = Force/Area

Example: The pressure exerted on the floor when by an individual wearing jogging shoes is much less than that exerted by an individual wearing high heels. The reason is that the force (weight) is the same, but the surface area of the heel of the jogging shoe is much larger than the surface area of a high heel.

Weight. The weight of an object on Earth is the gravitational force exerted on it by the earth. Weight is the force proportional to the mass of a body and the gravitational constant: weight = mass × the gravitational constant.

Example: If earth-calibrated spring scales were used on the moon, weights would be approximately one-sixth of what they are on Earth.

Mass. Mass is a measure of the quantity of matter in a body. On earth, weight is divided by the gravitational constant to obtain mass.

Resistance. Resistance in electrical terms is defined as the voltage divided by the current: resistance = voltage ÷ current.

Ohm's law. A basic law of physics that relates voltage and current to resistance, or for the blood flow situation, pressure and flow to peripheral resistance.

Force. Force in everyday language is a "push" or a "pull." When a grocery cart is pushed, a force is exerted on it. If a drawer is opened, a force is exerted on it.

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